Chapter 3

Sensors and Transducers

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INTRODUCTION

Terminology

According to Webster, a sensor is "any of various devices designed to detect, measure, or record physical phenomena, as radiation, heat, blood pressure, etc., and to respond, as by transmitting information, initiating changes, or operating controls." A transducer is "any of various devices that transmit energy from one system to another."[1] The number and type of transducers used by the medical profession is large and diversified. Physicians, nurses, respiratory therapists, and other clinicians caring for critically ill patients are confronted with some of the most sophisticated medical instrumentation in all of medical care. It seems that each year new methods and technologies are found for measuring parameters previously thought to be impossible or improbable.

The task of covering all the sensors and transducers that clinicians might encounter is daunting. A recent medical device encyclopedia further illustrated the size of the task[2]—it is comprised of four volumes and has more than 250 articles and nearly 2500 pages of text! From those articles, 33 are directly applicable to respiratory care, and include topics such as oxygen sensors, nitrogen analyzers, and colorimetry. More detailed background material and basic theory can also be found in the scientific literature and in texts on bioinstrumentation.[3–5]

For centuries, without medical instruments, physicians and other health care professionals relied on only their five senses (sight, hearing, smell, taste, and touch). Today's medical instruments use sensors and transducers and signal processing equipment to convert information about patients into a form that humans can perceive and understand. One of our primary concerns when making a measurement is to know its accuracy. The *accuracy* of a measured quantity is the true value minus the measured value divided by the true value (usually expressed as a percentage). True values are seldom known exactly, but a reference value traceable to the National Institute of Standards and Technology is usually used. The *precision* of a measurement expresses the number of distinguishable alternatives from which a given result is selected. A weight of 75.2 kg is more precise than a weight of 75 kg. High precision does not imply high

accuracy. *Resolution* is the smallest incremental quantity that can be measured with certainty. *Reproducibility* is the ability of an instrument to provide the same output for an equal input.

Medical instrumentation will have both *static* (pertaining to bodies or forces at rest or in equilibrium—opposite of dynamic) and *dynamic* (pertaining to dynamics; active—opposite of static) requirements. Static requirements usually refer to the performance of an instrument involving very low frequency events such as measurement of body weight. Dynamic requirements refer to dynamically changing measures that vary at high frequency, such as the arterial blood pressure or airflow from a rapidly breathing patient.

Instrumentation systems may have one or many of the components shown in Figure 3–1. The primary flow of information is from left to right. The physical quantity, property, or condition that the system measures is called the *signal*. Most medically important signals can be grouped into the following categories of biopotentials: electrocardiogram (ECG), pressure (blood pressure), flow (expiratory airflow), displacement (chest wall movement), impedance (chest wall bioimpedance), temperature (core body temperature), and chemical concentrations (K^+ or Po_2).

Transducers/Sensors

A transducer or sensor converts one form of energy to another. The final signal is usually electrical since the technology for amplifying and displaying electrical signals is well developed. The transducer should respond to only the form of energy present in the signal, to the exclusion

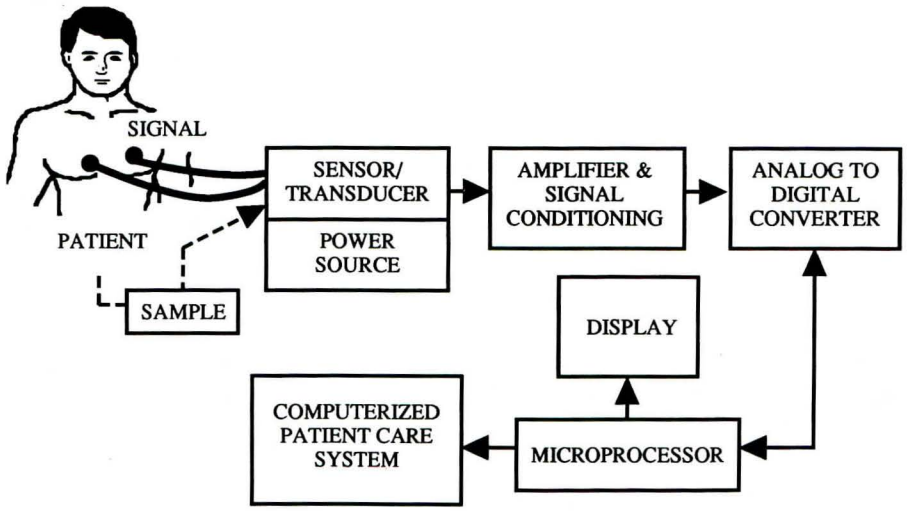

FIG 3–1.
Block diagram of a generalized instrumentation system. The transducer converts energy or information from the signal to another form of energy—usually electrical. The signal is then processed and displayed so that humans and computers can perceive the information.

of all others. In the real clinical situation this is seldom exactly possible, but can usually be accomplished well enough to provide useful information. The ideal transducer should interface with the patient in such a way that minimizes the energy extracted, while being minimally invasive. Many transducers have a primary sensing element such as a diaphragm, which converts pressure to displacement that is then sensed with variable resistive elements called strain gauges. Transducers can be something as conceptually simple as ECG electrodes. However, other transducers may require an energy source—for example, a dc voltage for a pressure gauge or pulses of electrical energy to activate the light sources in a pulse oximeter.

Amplifier and Signal Conditioning and Analog-to-Digital Conversion

Seldom can the output from a transducer be directly attached to the display device. Usually signals from transducers are small (in the milli- or microvolt range) and must be amplified and often filtered to remove unwanted signals. Most medical devices use only simple amplifiers and filters and then convert the desired signal to a digital form using an *analog-to-digital converter* (ADC). ADCs convert "analog" signals, such as an ECG, to digital form, usually with 10-bit (1 part in 1024) resolution at rates of from 100 to 300 times per second.

Ohm's law is a basic law of physics that relates voltage and current to resistance (analogous to the relationship between gas pressure, flow, and resistance). Figure 3–2 is a diagram of a simplified electrical/ mechanical circuit. The two types of current are dc (direct current) or

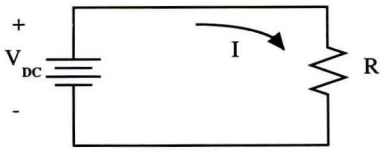

FIG 3–2.
Simplified diagram of an electrical/mechanical circuit showing the relationship of voltage and current to resistance (electrical) and pressure and flow to resistance (gas flow). Also the idea of conductance (the inverse of resistance) is presented.

$$\text{Voltage} = V = IR$$

where I = current and R =resistance.

$$\text{Pressure} = P = FR$$

where F = flow and R = resistance.

$$R = V/I \text{ (electrical) } R = P/F \text{ (gas flow)}$$

Conductance = $1/R$ = I/V (electrical) or = F/P (for gas flow).

battery circuits and ac (alternating current). In the United States, household current is 60 Hz (hertz, cycles per second). Figure 3–3 shows two simplified circuits with dc and ac excitation. Just as there is a resistance to dc flow, there is an electrical impedance related to capacitors and inductors. (Electrical impedance is defined as the apparent resistance in a circuit to the flow of alternating current, analogous to the actual resistance to a direct current.) Equations for the electrical impedances of the simplified ac circuits are shown in Figure 3–3.

Microprocessors

With microprocessors being inexpensive and reliable, the technology of signal processing, transmission, and display is accomplished with these devices. These microprocessors can take signals such as arterial blood pressure and in real time derive and display heart rate and systolic, diastolic, and mean blood pressures. These systems also compensate for undesirable transducer characteristics (e.g., nonlinearities) or they may average repetitive signals to reduce noise. Similar examples exist for respiratory care with ventilatory rate, mean airway pressure (MAP), and peak and plateau pressures.

Display and External Transmission

The results of the measurement process are displayed to the user in a form that is easily perceived. The usual forms of display are graphical plots, but display can also be in the form of digital results. For example, from a ventilator one might see the respiratory rate and tidal volume displayed. In recent years it has become even more important to transmit and store data in a computerized "charting" system so they can be shared with a diverse medical data user group. Typically, the clinical or blood gas laboratory is required to transmit data to the

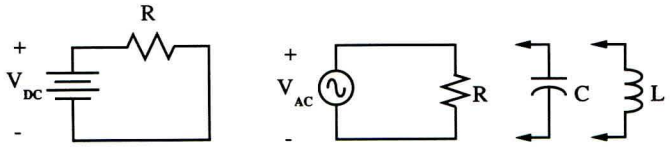

FIG 3–3.
Two simple dc and ac circuits. Note that AC circuits or circuits that have time-varying signals can have two additional electrical elements, capacitors (C) and inductors (L). These elements have "impedance factors" as shown.

$$X_C = \frac{1}{2\pi fc}$$

where f is the frequency in hertz.

$$X_L = 2\pi fL$$

intensive care unit (ICU). Just as the clinical staff in the ICU needs to know the laboratory results, they also need to know respiratory care charting results. In fact, the blood gas laboratory must know the F_{IO_2} and other ventilator parameters to make a reasonable interpretation of the blood gas results. With computerized expert systems becoming common, a high-priority task will be the sharing of respiratory care data.[6–13]

HOW COMMONLY USED TRANSDUCERS WORK

Pressure Transducers

Strain gauges are devices that allow measurement of a change in the dimensions, displacement, or deformation of an object. Electrical strain gauges are used extensively in pulmonary and cardiovascular monitoring to measure pressures, flows, and temperatures. Several types of electrical strain gauge pressure transducers are available: resistance, inductance, capacitance, and semiconductor transducers. Each of these pressure transducers has a diaphragm sensing element that deforms when pressure is applied. Depending on the type, the movement of the diaphragm is sensed by a change in resistance, inductance, or capacitance. The electrical resistance gauge is the most widely used. Recently, several pressure transducers have been made using electronic integrated circuit technology. These transducers are accurate, very reliable, inexpensive (< \$20), and disposable.[14–16] Figure 3–4 shows a diaphragm pressure transducer with resistive sensing elements attached. Figure 3–4, B shows, with the motion of the diaphragm greatly exaggerated, how an applied pressure causes the resistor wires to be stretched (increasing resistance) or relaxed (decreasing resistance). Volume displacements of pressure transducers used for measuring blood pressure are typically of the order of 0.1 mm^3 per 100 mm Hg applied. Thus the approximate displacement of the 4 × 4 mm pressure transducer diaphragm would be only 0.00625 mm with 100 mm Hg applied or 0.0000626 mm for each mm Hg applied!! This is indeed a very small displacement.

Figure 3–4, C, shows schematically how resistors are "etched" onto the surface of semiconductor materials using integrated circuit technology. The semiconductor "chip" becomes the diaphragm. Typically these semiconductor devices are very small (less than 4 × 4 mm).

Transducers for measuring blood pressure have recently been standardized by the American National Standards Institute (ANSI).[14] They have sensitivities of 5 μV per volt of excitation per mmHg pressure applied. Thus if 5 V of dc is applied, an output of 25 μV per mm Hg pressure applied is generated—a rather small signal. For measurements of airway pressures and pressures across Fleisch pneumotachometers, signal levels even smaller are produced, and errors caused by instabilities of pressure transducers and amplifiers become problematic.

Figure 3–5 shows a pressure transducer that measures the movement of the change in inductance by use of a moving magnetic armature attached to the diaphragm. This type of transducer is called a *linear variable inductance transducer* (LVDT), and it requires ac excitation. The resistive pressure transducer can be excited with either dc or ac voltage excitation.

Figure 3–6 shows a capacitance-type pressure transducer. Here the diaphragm moves the two plates of the capacitor closer together, causing the capacitance to increase. Capacitive transducers must also be energized with ac excitation.

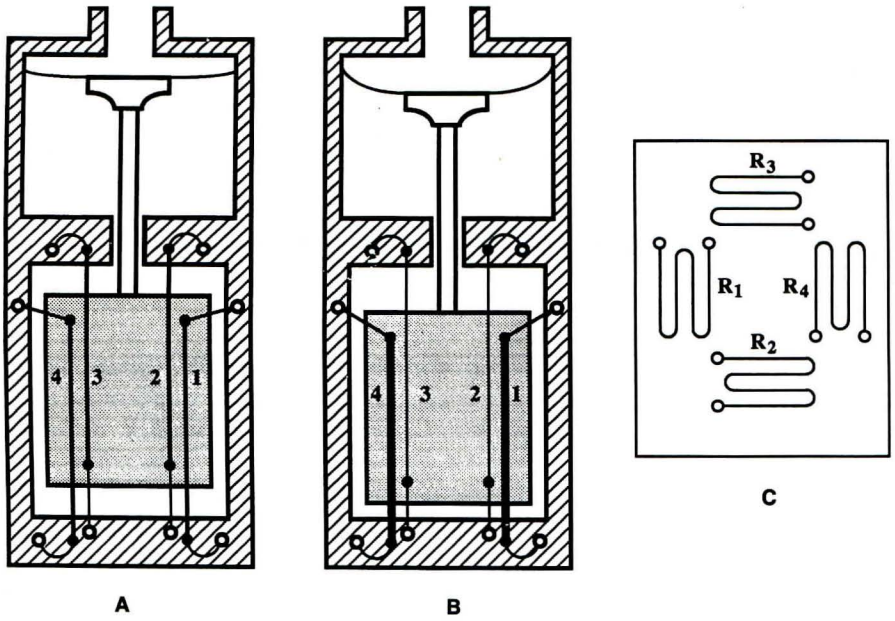

FIG 3–4.
Resistive strain gauge pressure transducer. Elements 1, 2, 3, and 4 are resistive wires. **A,** no pressure is applied. **B,** Pressure is applied to the transducer—the movement of the diaphragm is greatly exaggerated to illustrate the operating principle. **C,** Layout of resistors on a semiconductor "chip" for a modern disposable pressure transducer.

The principles used by electrical resistance strain gauges were discovered in 1856 by Lord Kelvin. He noted that the resistance of a metal wire increased with increasing strain and that different materials had different sensitivities to strain. Strain gauges usually use a Wheatstone bridge configuration (Fig 3–7), which is made up of four resistors that change resistance when a pressure is applied. Typically, two of the resistors increase their resistance and two decrease their resistance as shown in Figure 3–4, B. By configuring the transducer in this way, its ouput voltage is increased by four compared with a transducer where only one of the resistors changes with the pressure applied. The four active resistors in a typical pressure transducer are designed to have nearly the same resistance and are physically near each other. As a result, an unwanted change in "zero" or sensitivity is minimized because if the temperature increases slightly, each of the resistors increases its resistance proportionally, thus maintaining its "zero" point.

Temperature Transducers

Measurement of body temperature has been an important part of health care for centuries. Today it remains one of the simple, yet reliable, parameters measured in combination with other indicators to assess the state of health. For the respiratory and critical care clinician, the

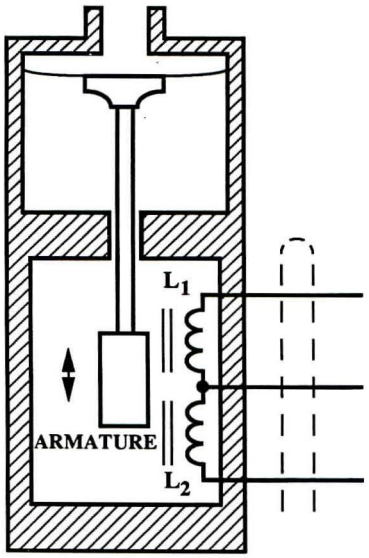

FIG 3–5.
LVDT (inductive) pressure transducer.

measurement of respiratory gas temperatures and body core temperatures to compensate for the temperature dependency of Po_2, Pco_2, and pH probes is essential.[17]

Each special application for monitoring temperature may have a unique need. For example, the classic mercury-in-glass thermometer may perform adequately for an occasional noninva-

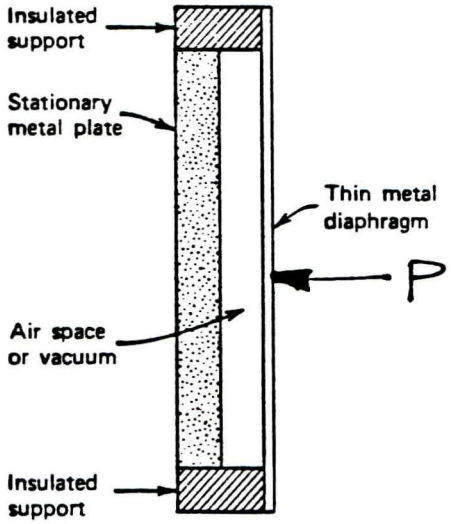

FIG 3–6.
Capacitance-type pressure transducer.

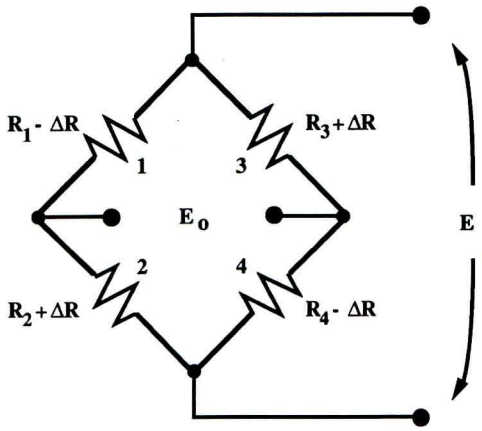

FIG 3–7.
Resistive pressure transducer configured as a Wheatstone bridge. Note from Figure 3–4 that elements 1 and 4 decrease in resistance (shorten) as pressure is applied, while elements 2 and 3 increase in resistance (elongate). The term *Eo* is the output voltage from the transducer and *E* is the excitation voltage.

sive measurement of body temperature, but these are not appropriate where size, speed of response, and breakage are considerations. Thermocouples or thermistor sensors are convenient electronic substitutes that offer small size and fast response at relatively low cost.

Thermistors

A thermistor is small in size and has a high resolution and rapid response for blood temperature measurement at a point along an indwelling catheter (e.g., thermodilution cardiac output monitoring with a Swan-Ganz catheter). A thermistor sensor is fabricated by forming a powdered semiconductor material (usually a metal oxide) into a small bead around two lead wires, or by sintering the semiconductor into a pellet shape, then coating its opposite faces with conducting electrodes.[17] The resistance of the resulting thermistor has a large variation with temperature. Unfortunately, the resistance-temperature curve is nonlinear (Fig 3–8). However, the unique and stable characteristics of thermistors are easily determined. Thermistors can be manufactured to very small sizes (of the order of 0.2 mm in diameter), to have fast response times (<0.1 seconds), and to have a temperature range of 100°C. They can be pretrimmed to be interchangeable within ±0.1°C of each other. Thermistors have a much higher sensitivity than thermocouples and thus have better resolution. Their primary disadvantage is their nonlinearity.

Thermocouples

When two dissimilar metals such as copper and constantan are placed in contact with each other, a small voltage can be measured across the metallic contact junction.[17] This voltage varies with temperature and can be used to measure body temperatures. For the biological temperature range, the combination of copper and constantan (an alloy of 55% copper and

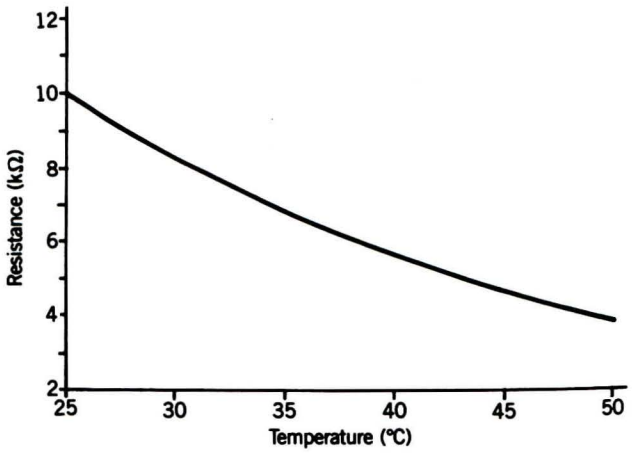

FIG 3–8.
Resistance versus temperature for a thermistor pellet that has a nominal resistance of 10 kΩ at 25°C. (*From Christensen DA: Thermometry, in Webster J (ed):* Encyclopedia of Medical Devices and Instrumentation. *New York, Wiley-Interscience, 1988, Vol 4, pp. 2759–2765. Used by permission.*)

45% nickel) is appropriate. These are T-type thermocouples and have a sensitivity of about 43 μV/°C. They have good linearity over the temperature range of 25 to 50°C, they are inexpensive, and they have good stability.

Spectrophotometry/Colorimetry

Colorimetry is based on the observation that molecules absorb light. The wavelengths and efficiency of absorption depend on both the structure of the molecule and its environment, making absorption of light a useful tool for characterizing both small molecules and macromolecules such as hemoglobin. Colorimetric methods use these light-absorbing spectral properties to measure the presence and concentration of substances.[18] Figure 3–9 is a block diagram of a spectrophotometer/colorimeter. Figure 3–10 shows the almost linear absorption of infrared radiation at 4.26 μm when shined through a 4-mm-thick sample of CO_2.

The probability of light absorption at a single wavelength is described by the Beer-Lambert law. The passage of light through any given thickness of any substance results in absorption of a constant fraction of the incident light. In differential equation form this is stated as:

$$dI/I = -KC\,dL$$

where dI/I is the fraction of light absorbed by a layer of thickness dL,K is a constant that depends on the properties of the substance, and C is the concentration of the absorbing substance. Integrating the above equation yields:

$$\ln(I_o/I) = KCL$$

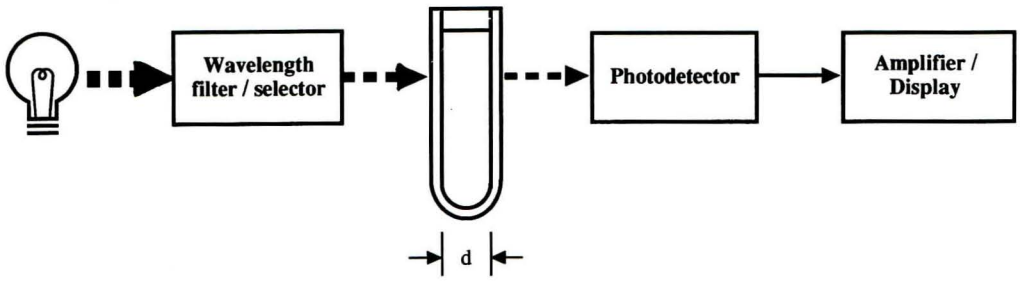

FIG 3–9.
Block diagram of a spectrophotometer/colorimeter.

where ln is the natural logarithm, L is the path length, I_0 is the initial intensity, and I is the final intensity of light after passing through the sample. The quantity (I/I_0) is known as the transmittance (T). Optical density (OD) (also called absorbance) is related to T by the equation:

$$\text{OD} = -\log T = \text{Absorbance}$$

A plot of OD as a function of concentration is ideally a straight line.

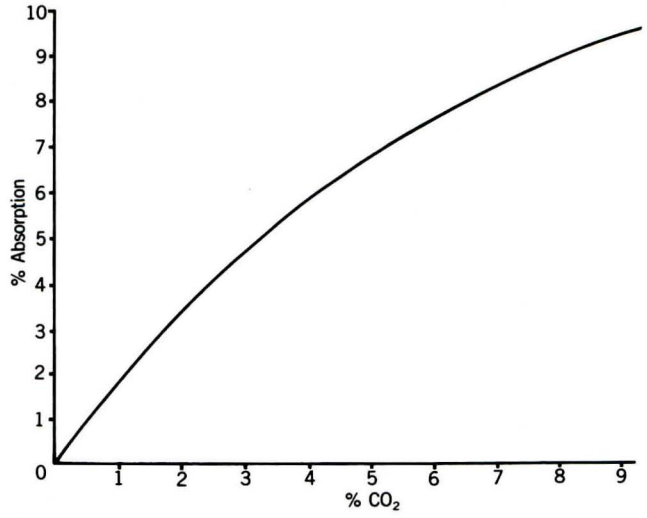

FIG 3–10.
Absorbance versus concentration for infrared radiation shined through a cuvette of CO_2. (*Ref. [19]*)

Blood Gas Analyzers

pH Electrode

The measurement of pH is accomplished by using a special glass electrode that generates an electrical potential when solutions of differing pH are placed on the two sides of its membrane. pH is measured in virtually every blood gas analyzer using these glass electrodes (Fig 3–11). The glass used in the electrode is a member of the class of ion-specific electrodes that react only with a specific ion. The pH glass sensor, introduced in the early 1900s, was the first example of an ion-selective electrode. The Nernst equation that follows applies to this thin ion-specific glass. Therefore, the voltage across the membrane (E) changes by 61.5 mV/pH unit at body temperature (37°C). Since the range of physiological pH is only about 0.6 pH units, the change in voltage for the entire pH range is only 36.9 mV. The pH meter must be capable of accurately measuring changes of about 0.1 mV. In typical blood gas analyzers, the pH of very small samples (20 μL) can be measured.

The Nernst equation for the hydrogen ion (H^+) is:

$$EH^+ = (RT/F) \cdot \ln [(H^+{}_o)/(H^+ i)]$$

where R is the gas constant, F is Faraday's constant, and T is the absolute temperature (kelvin).

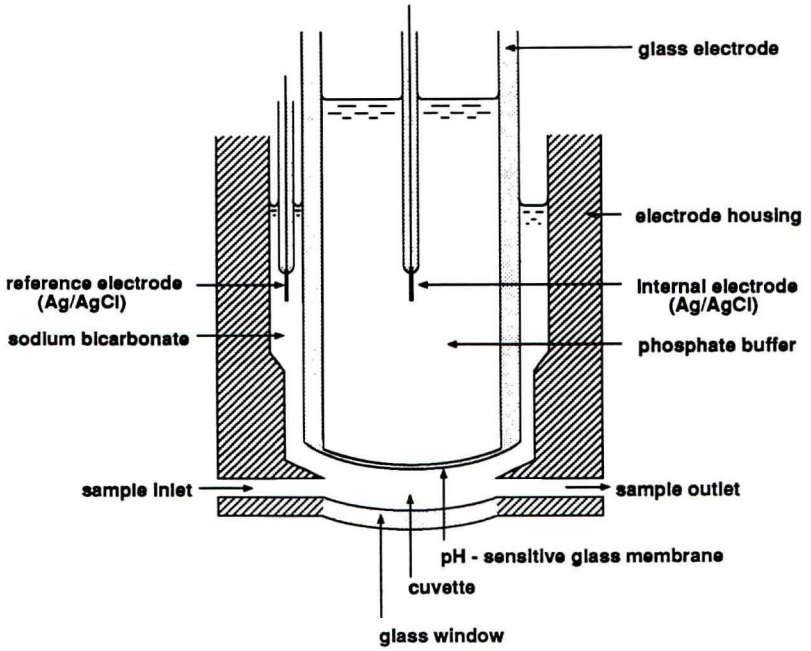

FIG 3–11.
pH electrode.

At body temperature (37°C) the quantity (RT/F) is 61.5 mV. Therefore, at body temperature the above equation becomes:

$$EH^+ = 0.0615 \cdot \log_{10} [H^+]$$

where $\log_{10} [H^+]$ is defined as a pH unit.

P_{CO_2} Electrode

The CO_2 electrode was first described by Stow and his colleagues [19] in 1957, and was later improved by Severinghaus and Bradley.[20, 21] The basic idea of the electrode is to allow an unknown CO_2 sample to equilibrate with an aqueous solution, and then measure the pH. Figure 3–12 shows that a P_{CO_2} electrode consists of a glass pH electrode covered with a Teflon or Silastic membrane. A thin layer of water containing salt and bicarbonate ion is held between the glass and the Teflon by a spacer (nylon stocking or filter paper are also used as spacers).

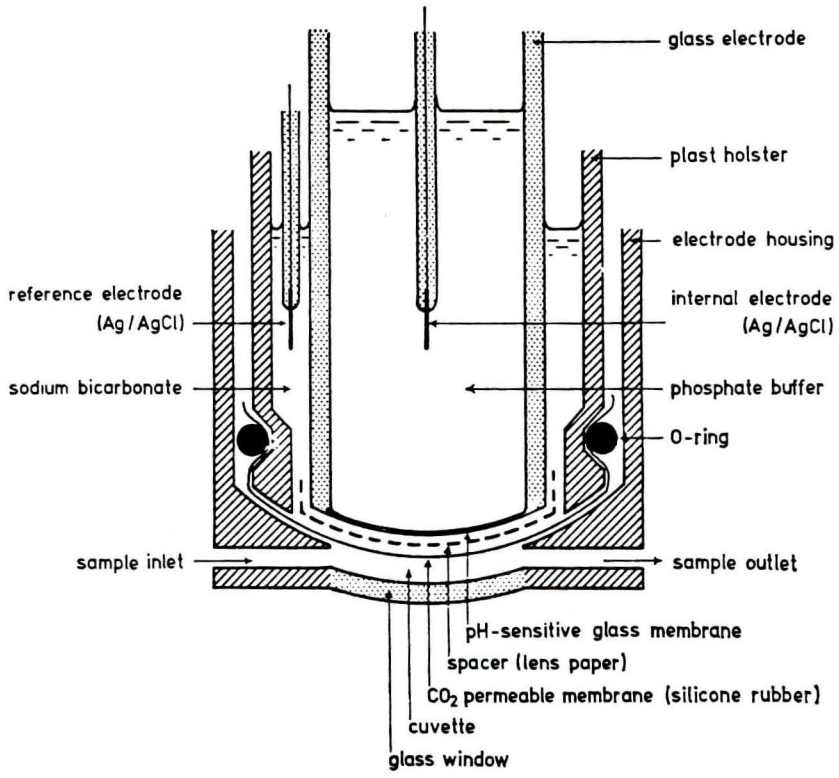

FIG 3–12.
Severinghaus P_{CO_2} electrode. With earlier electrodes the spacer element, shown here as lens paper, was actually a nylon "gauze" made from a nylon hose worn by women.

The CO_2 gas molecules from the sample diffuse through the Teflon membrane and react with water to form hydrogen ions and bicarbonate:

$$CO_2 + H_2O \rightarrow H_2CO_3 \rightarrow H^+ + HCO_3^-$$

Using a reference electrode in contact with the water film permits the measurement of the resulting pH. With low-bicarbonate concentrations, the measured pH is then determined by the Henderson-Hasselbalch equation:

$$pH = pK + \log([HCO_3^-]/[CO_2])$$

or in simplified form:

$$pH = C - \log[Pco_2]$$

The electrode senses a pH change of 1 pH unit for a tenfold change in Pco_2. The Pco_2 can be measured over the range of 10 to 90 mm Hg—the range of clinical interest.

PO2 Electrode

Oxygen was first measured in 1774 by the French chemist Antoine-Laurent Lavoisier. More than 100 years later, Danneel, working in Walther Nernst's laboratory, showed that dissolved oxygen reacted with electrodes in solution when a voltage was applied between the electrodes. However, it was not until 1925 that Jaroslav Heyrovski made the application of oxygen polarography practical, for which he earned the 1959 Nobel prize.[22, 23] Figure 3–13 shows the basic Leland Clark electrode introduced in 1956 on which the designs of most contemporary Po_2 electrodes are based. Clark had the idea of covering the sensor with a membrane to increase the stability of Po_2 readings and to minimize the effects of fluid flow past the electrode. The Clark electrode consists of a tiny (25-μm) platinum wire sealed in glass with the tip exposed by polishing. The tip is covered with an oxygen-permeable membrane, usually polypropylene. The Clark polarographic sensor, with about − 0.7 V applied, gives a current output that is almost exactly linearly related to Po_2. The amount of oxygen extracted from the sample is very small, so stirring of the sample is not required.[24] The polarographic electrode must be calibrated to establish linearity, drift, response time, zero current level, and flow sensitivities. Most often this is done by using 100% oxygen, room air, and a known gas mixture.[23] The oxygen electrode can be used with oxygen in a gaseous or liquid phase and does not require equilibration with a blood sample. The output current of the Po_2 electrode is linear within ± 1% from 0 to 100% oxygen. At sea level this is a range from 0 to 760 mm Hg.

Severinghaus and Astrup, two of the pioneers of blood gas analysis, have written a historical review of blood gas instrumentation development; the reader is referred to this excellent series.[25–31]

CO Oximetry

The basis of oximetry of multiple hemoglobin measures is dependent on the optical absorption spectra of hemoglobin (Hb), carboxy (carbon monoxide) hemoglobin (COHb), and methemo-

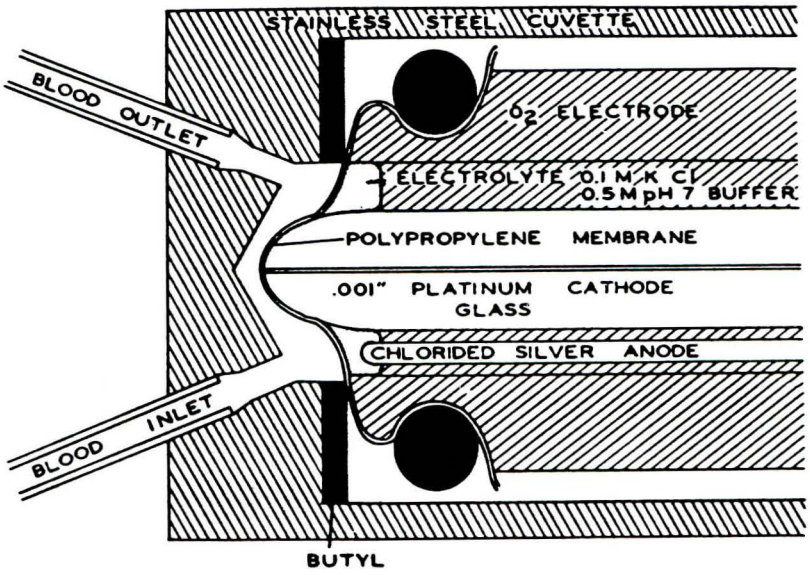

FIG 3–13.
Clark PO_2 electrode. (*From Severinghaus JW: Blood gas concentrations in* Handbook of Physiology. *Bethesda, MD, American Physiological Society, 1965, Sec 3, Vol 2, Chap 61, pp 1475–1487. Used by permission.*)

globin (MetHb). By measuring the optical absorption at multiple wavelengths, all of the above-mentioned parameters can be measured.

Continuous Oximetry

The basis for continuous So_2 measurement relies on the difference in optical absorption spectra of Hb and Hbo_2 (Fig 3–14). As a result, various optical methods for measuring oxygen saturation have been developed using two or more wavelengths in the near-infrared region: LAMBA1, where the largest difference in light absorption between Hb and Hbo_2 exists (about 660 nm), and LAMBDA2 at the wavelength where the two hemoglobins have the same absorbance (805 nm). The 805-nm wavelength, where the two hemoglobins have the same absorbance, is called the *isosbestic wavelength* (Fig 3–14). This principle is used to measure hemoglobin concentrations, oxygen saturation, and the saturation of carboxyhemoglobin and methemoglobin in laboratory blood gas analyzers. Modifications and engineering extrapolation of these principles are applied to the pulse oximeter.

Fiber-Optic Reflection Oximeters

The first report[29, 32] of the measurement of oxygen saturation by reflected light from blood was described in 1949. It was not until fiber-optic catheters and inexpensive light sources and

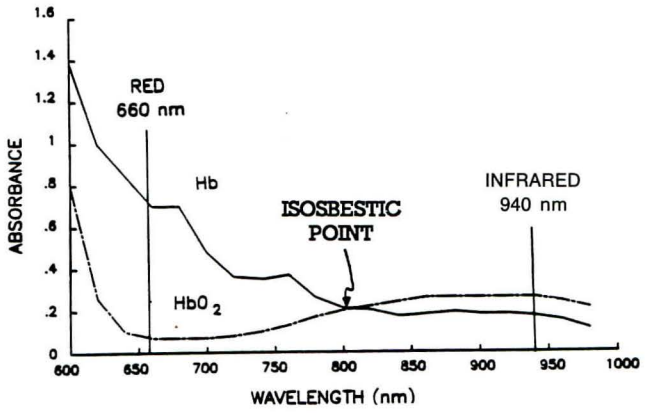

FIG 3–14.
Optical absorption spectra of Hb, Hbo_2, and Hbco in the visible and near-infrared wavelength regions. (*From Gardner RM: Pulse oximetry: Is it monitoring's "silver bullet"?* J Cardiovasc Nurs *1987; 1:79–83. Used by permission.*)

detectors became available that this mode of measuring oxygen saturation became widely used.[33] Mixed venous oxygen saturation measurements are now made for selected patients with disposable fiberoptic Swan-Ganz catheters using this technology.[34]

Pulse Oximeters

A recent and major advancement in oximetry has been accomplished with the pulse oximeter.[31, 35–46] Contemporary pulse oximeters use two light-emitting diodes (LEDs) to shine light through a variety of body locations (fingers, ears, or nose) where a small photodiode detects the transmitted signal (See Chapter 10—Pulse Oximetry). These inexpensive disposable probes and their associated processing and display capability have revolutionized oxygen saturation monitoring.[31, 36]

Figure 3–15 shows a block diagram of a pulse oximeter. The specific red and infrared wavelengths of light are generated by LEDs. These LEDs are ubiquitous, being used first in digital watches and now often used for displays on intravenous pumps. The receiving photodiode detects the light from the pulsations of the red and infrared light sources after they have been transmitted and absorbed by the tissue. A typical signal generated by one of the pulsating light sources is shown in Figure 3–16. It has been observed that when the pulsatile components of the light are less than 0.5% of the steady (dc) component, the system's accuracy falls.

Oxygen saturation can now be measured noninvasively between 50 to 100% with an accuracy of about 2.5% (ISD). Pulse oximeters do not, however, measure other types of hemoglobin such as carboxyhemoglobin or methemoglobin. Pulse oximetry has revitalized oximetry, which is a reversal of a trend started when the oxygen electrode became available and Po_2 measurements became feasible. Severinghaus and Astrup have stated "Pulse oximetry is arguably the most significant technologic advance ever made in monitoring the well-being and safety of patients during anesthesia, recovery, and critical care."[31]

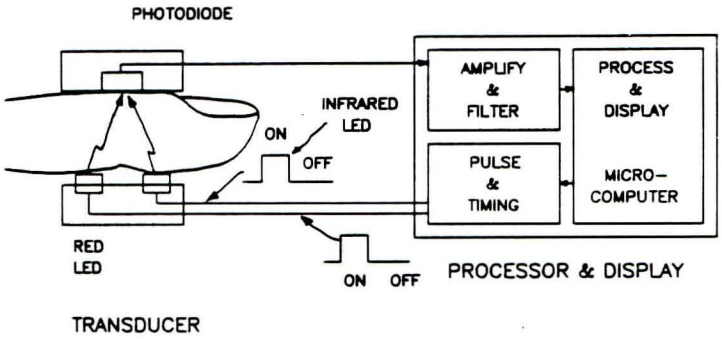

FIG 3–15.
Schematic block diagram of a pulse oximeter showing the transducer with its red and infrared LEDs and the photodiode detector. Each LED is pulsed several times each second, and the pulses are received by the photodiode. The signals from the photodiode are processed by the microcomputer and shown on a display. (*From Gardner RM: Pulse oximetry: Is it monitoring's "silver bullet"?* J Cardiovasc Nurs *1987; 1:79–83. Used by permission.*)

Continuous Respiratory Gas Measurement

Two types of gas sampling techniques are used to analyze respiratory gases from a patient's airway circuit: mainstream (in line) and sidestream (diverting) (See Chapter 12—Capnography). Mainstream sampling uses a transducer housed in an airway connector placed "in line"

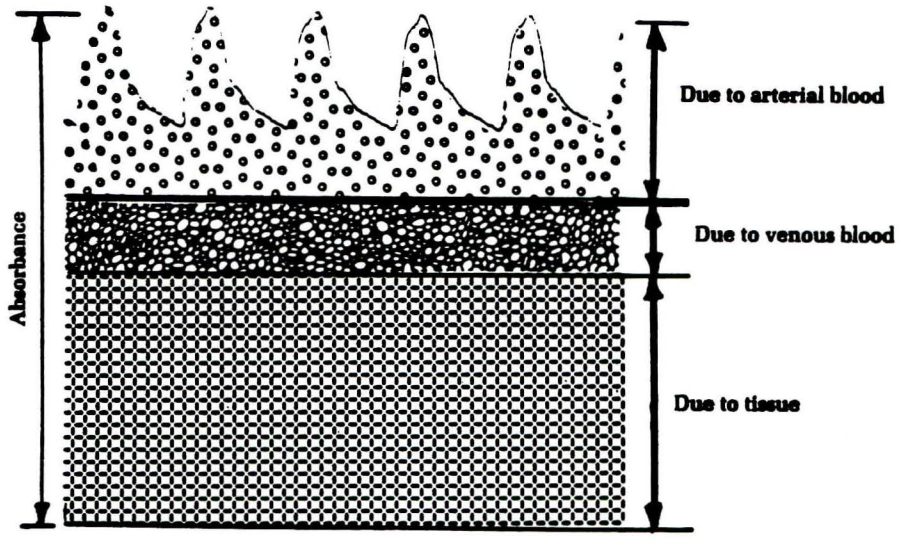

FIG 3–16.
Pulsatile light output from the photodiode generated from one of the LED input signals. The typical plethysmographic signal is superimposed on the nonspecific absorption signals. The pulsatile signal is processed to give nearly beat-by-beat oxygen saturation. (*From Gardner RM: Pulse oximetry: Is it monitoring's "silver bullet?"* J Cardiovasc Nurs *1987; 1:79–83. Used by permission.*)

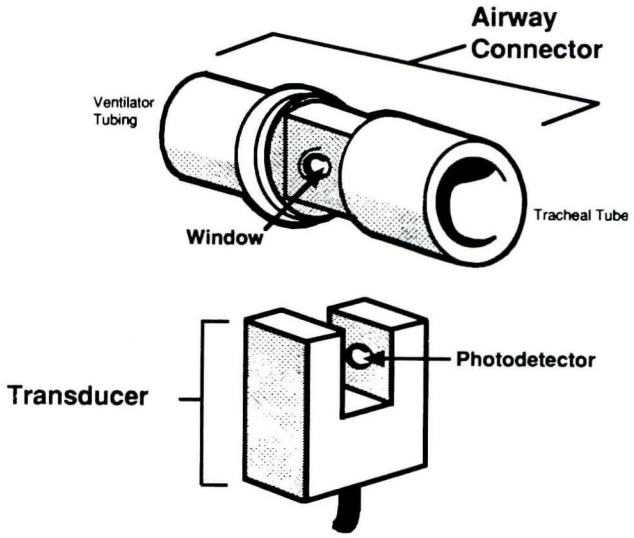

FIG 3–17.
Schematic of in-line continuous mainstream infrared CO_2 gas analysis method. (*From Szaflarski NL, Cohen NH: Use of capnography in critically ill adults.* Heart Lung *1991; 20:363–374. Used by permission.*)

with the patient's breathing circuit. Figure 3–17 is an illustration of an in-line infrared CO_2 measurement system. Sidestream or diverting sampling systems actively withdraw gas from the patient's airway via a sampling tube to the analyzer. Figure 3–18 shows a schematic of a sidestream sampling method. For the sidestream method, a small internal diameter tube must be used to allow for rapid gas withdrawal,[47] and gas analysis takes place in an external sensor. Because the gas sample must be transported to the external sensor, unavoidable transit times

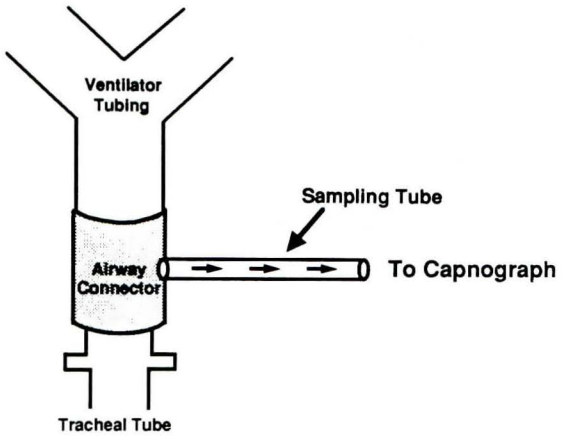

FIG 3–18.
Schematic of sidestream continuous gas sampling method. (*From Szaflarski NL, Cohen NH: Use of capnography in critically ill adults.* Heart & Lung *1991; 20:363–374. Used by permission.*)

of as long as a few seconds may occur. Also, gas sampling requires that water vapor and airway secretions not interfere with the flow of gases to the external transducer.

CO_2 Gas Analyzers

CO_2 can be measured in its gaseous form and in blood and tissues. The measurement technology is usually different for each measurement. In its gaseous form, CO_2 is usually measured in a respiratory circuit using infrared absorption, while measurements of P_{CO_2} in blood are usually made with an electrode system. Since the two use rather different methodologies and sensors, each will be discussed.

Just as P_{CO_2} electrodes can be used in blood gas machines, they can also be used to measure respiratory gases. Because the response time of the CO_2 electrodes is slow and because the electrodes are somewhat temperamental, they are seldom used for continuous gas analysis.

Various gases and liquids absorb power (energy) in specific regions of the electromagnetic spectrum, as has already been discussed for spectroscopy and colorimetry. Figure 3–19 shows that CO_2 has an infrared absorption peak at 4.26 μm, which is almost clear of other potentially interfering gases such as nitrogen, oxygen, nitrous oxide, halothane, and water.[21] The relationship between $\%CO_2$ and optical density (absorption) is almost linear. The technical tasks that must be solved to make CO_2 gas sensors successful are (1) choosing the appropriate pathlength to give the best compromise of sensitivity and linearity, (2) choosing an appropriate infrared light emitter—a task greatly simplified because of the need to develop infrared emitters for transoceanic fiber-optic communications, and (3) choosing an appropriate detector to sense the infrared signal—again a technology that is now within reason because of the semiconductor sensor revolution and the need for better worldwide optical fiber communications.[21]

As a result of emitter and transducer developments, the ability to measure CO_2 in line in ventilators is becoming common practice. Previous sensors required the use of a rotating "chopper wheel" that had multiple filters and reference gases on its circumference. Because it is now possible to measure CO_2 at 4.26 μm and water at 1.45 μm, the measurement of respiratory gases is now possible under normal clinical situations.

O_2 Gas Analyzers

Continuous O_2 analyzers have long been used in breathing circuits to ascertain that patients were receiving the desired level of oxygen therapy. The polarographic oxygen sensor (P_{O_2} electrode) is the most common gas-composition sensor used in respiratory care. The polarographic P_{O_2} electrode, such as that used for continuous ventilator gas analysis, is much the same as the sensor used in blood gas analyzers. The response time of these sensors is usually in the range of 30 to 360 seconds, depending on the membrane uses.

Some oxygen sensors such as the yttria-stabilized zirconia electrode, illustrated in Figure 3–20, incorporate a solid electrolyte.[48, 49] The zirconia crystals demonstrate oxygen-ion conduction at temperatures above 600°C. The crystal has a thin layer of porous platinum on each side that forms the electrodes. When the sample gas has a different concentration than the reference gas, a voltage difference (V) is generated across the crystal. The equation for the potential is:

$$V = \frac{RT}{4F} \cdot \ln \left[\frac{P_{(\text{sample})}\,O_2}{P_{(\text{reference})}\,O_2} \right]$$

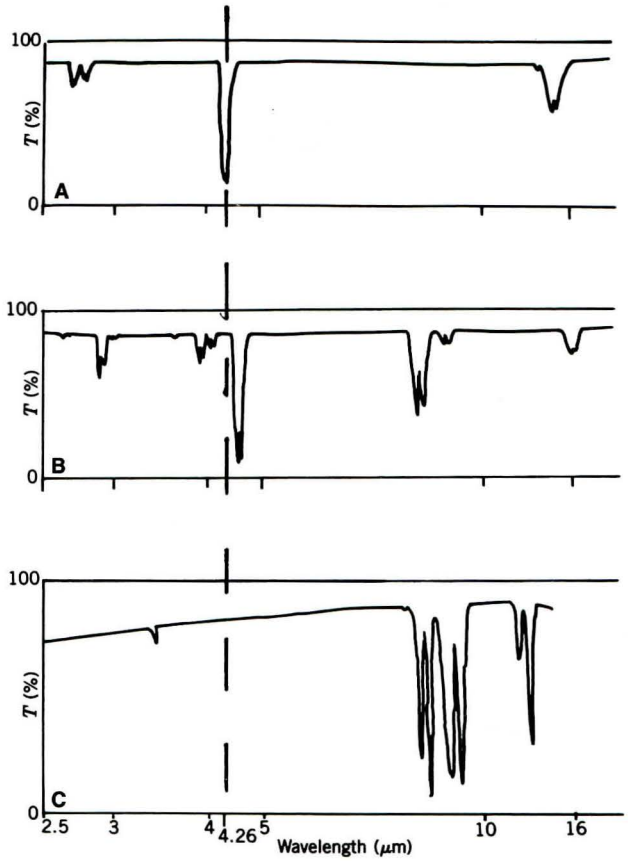

FIG 3–19.
Transmission spectrums in the infrared region for **A,** CO_2, **B,** nitrous oxide, and **C,** halothane. Note that CO_2 has a good absorption peak at 4.26 μm that is independent of the other two anesthetic gases. (*From Coombes RG, Halsall D: Carbon dioxide analyzers, in Webster J (ed):* Encyclopedia of Medical Devices and Instrumentation. *New York, Wiley-Interscience, 1988, Vol 1, pp. 556–569. Used by permission.*)

where F is the Faraday constant and T is the operating temperature of the cell (kelvin).

Fuel or galvanic cells are another large class of commonly used oxygen sensors.[49] These devices require no external power and are essentially batteries whose output current is a function of the Po_2. A diagram of the fuel cell sensor is shown in Figure 3–21. The fuel cell usually has a gold cathode and a lead anode immersed in potassium hydroxide (KOH) electrolyte. The reactions are:

$$O_2 + 2H_2O + 4e^- \rightarrow 2H_2O_2 + 4e^- \rightarrow 4OH^- \quad \text{(cathode)}$$
$$2Pb + 6OH^- \rightarrow 2PbO_2H^- + 2H_2O + 4e^- \quad \text{(anode)}$$

The oxidation of lead to lead hydroxide at the anode provides the four electrons necessary for

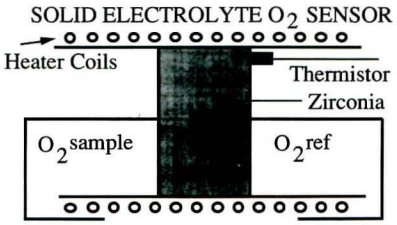

FIG 3–20.
Zirconium cell solid-state oxygen sensor. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.*)

the reduction of oxygen at the cathode. Because the lead and hydroxide are consumed in the process, fuel cells have a finite life of about three months when exposed to 100% oxygen and about 15 months when exposed to room air.

The paramagnetic method makes use of a principle of oxygen described by Linus Pauling (it is sometimes called the *Pauling principle*). Oxygen exhibits an unusual attraction to an applied magnetic field—a phenomenon known as *paramagnetic susceptibility*. Figure 3–22 illustrates the basic function of a paramagnetic analyzers.[48, 49] A reference gas is passed through two symmetrical pathways joined in the middle by a differential pressure transducer. A strong pulsating electromagnetic field (EMF) is placed on one pathway just before the gas outlet, and the sample gas is injected at the junction of the two pathways near the outlet. If the sample gas has a higher paramagnetic affinity than the reference gas, it will be attracted to the magnetic field and produce a restricted flow in that pathway. This restriction produces a pressure gradient between the pathway that is proportional to the difference in Po_2 between the reference and the sample gas. The paramagnetic sensor is fast, having a response time of less than 200 ms. As a result, the sensor is ideally suited for fast, breath-to-breath applications.

Multiple Gas Analyzers

We often need to measure multiple gases from patients. For example, in anesthesiology it is important to measure oxygen, carbon dioxide, nitrogen, and the anesthetic gases.[50] Other

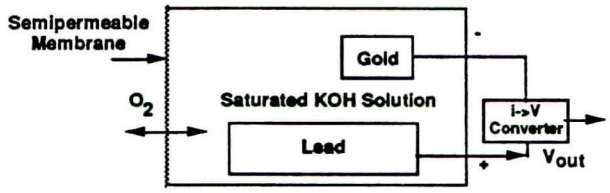

FIG 3–21.
Fuel cell oxygen sensor. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.*)

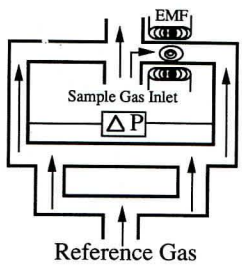

FIG 3–22.
Paramagnetic oxygen sensor. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.*)

applications in pulmonary function laboratories and ICUs have also been described. Most of the multiple gas analyzers currently available use a sidestream sampling method.

A mass spectrometer is a device that separates a stream of charged particles (ions) into a spectrum according to their mass-to-charge ratios and determines the relative abundance of each type of ion present.[51] Medical mass spectrometers have been used extensively in pulmonary medicine. Figure 3–23 shows the elements of such a mass spectrometer. These devices are "leaky" vacuum tubes that usually take a sample of gas from ventilator or anesthesia circuits. The sample inlet has a capillary tube that is often made of Teflon. Gas is drawn through this tube by a vacuum pump. A small amount of gas in the inlet chamber "leaks" into the ionization chamber, where a stream of electrons bombards the sample and causes the molecules to lose electrons and become positive ions. These positive ions are then focused and accelerated with electrical fields into the strong magnetic field of the dispersion chamber. The dispersion chamber is also fitted with collectors. The molecules measured are collected by appropriately placed collectors. A current proportional to the number of molecules collected is the fraction of the gas mixture the molecule represents.

Mass spectrometers are used in pulmonary function laboratories, anesthesia, some ICUs, and for continuous blood gas analysis. They have the distinct advantage that they can synchronously measure multiple respiratory gases and trace gases to enable measurements of complex pulmonary function in the clinical pulmonary function laboratory and during anesthesia.[52–57] Mass spectrometers have gained wide use in anesthesia monitoring. They help the anesthesiologist determine if the anesthesia delivery system is functioning correctly, the level of anesthesia, the patient uptake of anesthetic gas, and some indication of the physiological status of the patient. Multiplexed mass spectrometers (Fig 3–24) are used for anesthetic gases in surgery and in some ICUs.

Gas chromatography is a method of separating gas mixtures into their various components by forcing them through a long, narrow column packed with either solid or liquid material.[48] *Chromatography* was the term originally given to this method because substances were identified by the color of the reaction with the material lining the column. However, more sophisticated and diverse methods of separation are now used, and most do not rely on color to identify the substance.

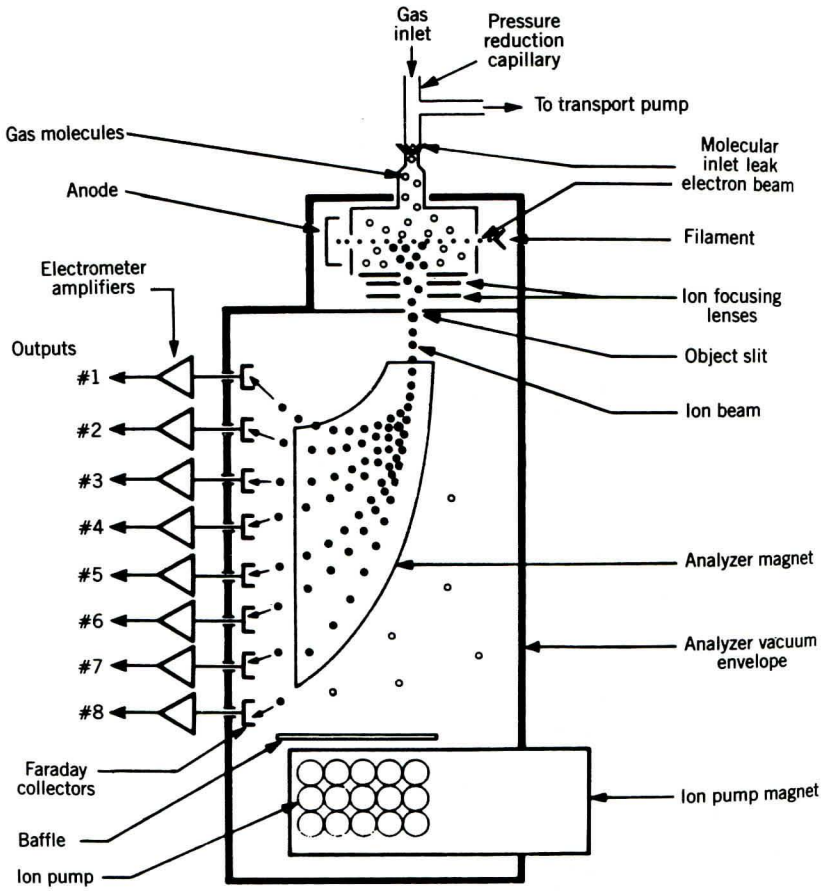

FIG 3–23.
Magnetic sector-type mass spectrometer showing its ion pump and vacuum chamber. (*From Sodal IE, Clark JS, Swanson GD: Mass spectrometers in medical monitoring, in Webster J (ed):* Encyclopedia of Medical Devices and Instrumentation. *New York, Wiley-Interscience, 1988, Vol 3, pp. 1848–1859. Used by permission.*)

A gas chromatograph consists of a sample port leading to a chamber where the sample is heated.[48] Heating removes the problem of water vapor in the sample. The sample is then added to a known concentration of a reference gas, usually helium for respiratory measurements. The sampled gas is forced by a pump through a long, narrow column. The material in the column separates the gases based on their molecular weight and viscosities. Each gas arrives at the detector at a distinct time (Figure 3–25). A peak is seen for each gas as it reaches the detector. The area under the peaks is proportional to the gas concentration. An advantage of the gas chromatograph is that only small samples are needed. However, the analysis takes from 15 minutes to 2 hours. Thus the method is not applicable for breath-by-breath analysis.

Raman spectroscopy is a versatile method of molecular analysis first postulated by Adolf

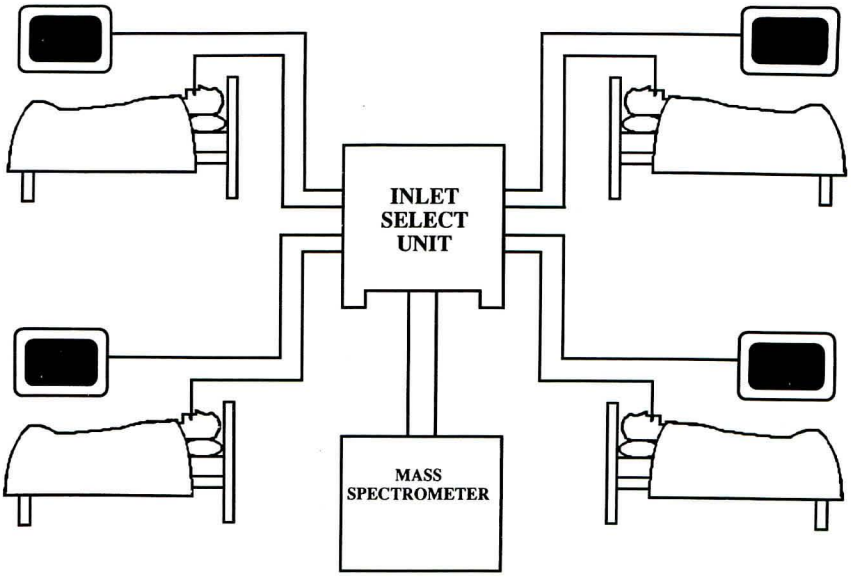

FIG 3–24.
A multiplexed mass spectrometer patient monitoring system that services four patients.

Smekal in 1923, and first observed by Sir Chandrasekhara Venkata Raman in 1928.[48, 58–60] When light collides with gas molecules, the photon loses energy to the gas molecule and subsequently has less energy and, consequently, a longer wavelength (lower frequency). The magnitude of this wavelength shift corresponds to the vibrational and rotational energies of

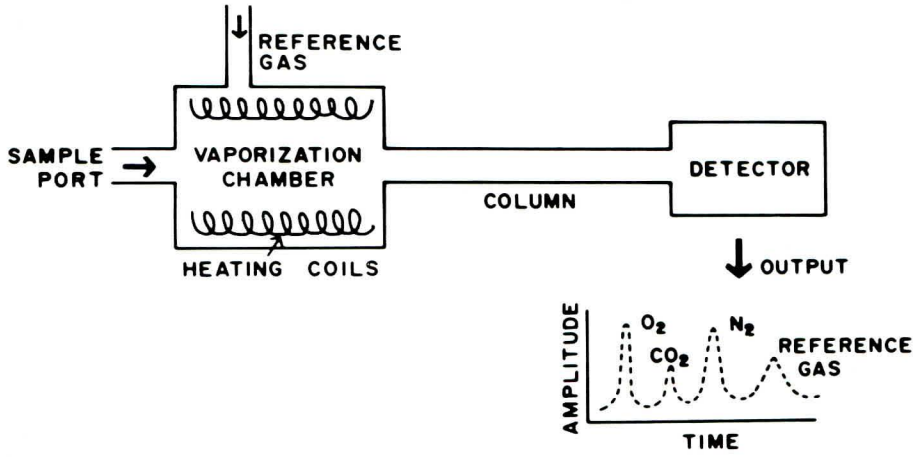

FIG 3–25.
Gas chromatograph multiple gas analyzer. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.*)

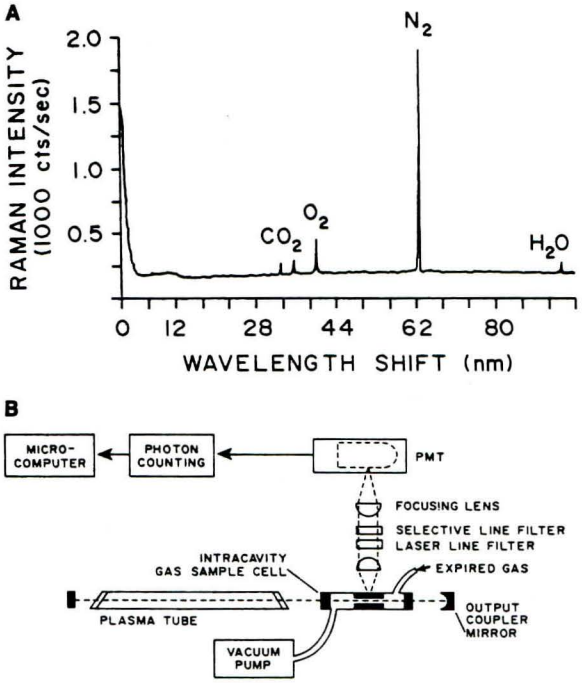

FIG 3–26.
A, Raman scatter spectrum for a typical sample of expired respiratory gas. **B,** System for Raman spectrographic analysis of expired respiratory gases. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.*)

the molecules and is different for each gas. A Raman spectrum for a typical expired sample of respiratory gas is shown[48, 58] in Figure 3–26. The amplitudes of the peaks are proportional to the concentration of the gas present. Gas concentrations are measured by isolating the specific characteristic Raman wavelengths for each of the gases by using multiple optical wavelength filters. The magnitude of the Raman signal is small, but by using high-intensity light sources such as lasers, the signal levels are good and thus low concentrations of gases such as anesthetic agents can be measured.[59] The Raman scattering takes place almost instantaneously (about 10 picoseconds) so that breath-to-breath measurements are possible—they are limited only by the dynamics of the sidestream sampling mechanism.

Using a sidestream sampling technique, a multigas monitor based on acoustic principles was recently introduced.[48] The device measures all gases and vapor concentrations except oxygen by a special infrared absorption technique called *photoacoustic spectroscopy*. A precision microphone detects the energy the gases absorb; it "listens" as they expand and contract. Oxygen concentration is determined by magnetoacoustic techniques. The manufacturer (Brüel & Kajaer) claims that the sensing system requires less frequent calibration and that it is more accurate than more conventional infrared techniques.

Gas Flow/Volume

Gas Flow

Various gas flowmeters are available: rotameters, ultrasonic and thermal flowmeters, and pneumotachographs. Rotameter-type flowmeters place a small turbine in the flow path. For steady flows with uniform gas composition, this type of flowmeter can be calibrated to be very accurate. Unfortunately, with the rapidly varying flows and variable gas compositions found in respiratory measurements, this type of flowmeter is often too inaccurate to be useful. Ultrasonic flowmeters have also been used for gas flow, but they also suffer from limitations similar to the rotameter. Thermal flowmeters employ a sensing elements such as metal wires, metal films, and thermistors whose resistance changes with temperature. Flowmeters that use a single, temperature-compensated, heated wire (the so-called "hot-wire" anemometer) with linearizing circuits provide good unidirectional flow indications. Respiratory gas flowing in either direction "cools" the wire, which in turn requires more energy to heat it and maintain a constant temperature. The unidirectional flow measuring capability is a major limitation of the hot-wire anemometer.[3, 61]

One of the most popular and accurate respiratory flowmeters is the 1925 Fleisch pneumotachograph, named after its inventor.[62] This device depends on the measurement of a pressure drop across a flow resistor. The flow resistor is usually either a screen or a series of capillary tubes in parallel. Figure 3–27 shows a Fleisch pneumotachometer with a screen and parallel tubes in series. This fixed resistor in the flow path causes a pressure drop that is (nearly) linearly

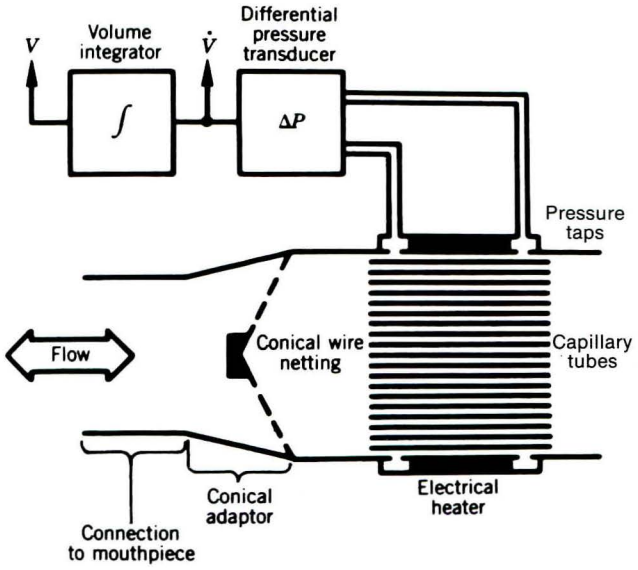

FIG 3–27.
Fleisch pneumotachometer. A linear resistance-flow element.

related to flow. The equation for Poiseuille's law describes the pressure-flow characteristic of the Fleisch pneumotachograph:

$$P_1 - P_2 = \frac{8uLF}{\pi r^4}$$

where $P_1 - P_2$ is the pressure drop across the flow head, u is the gas viscosity, L is the length of the device, r is the radius of the device, and F is the flow. A differential pressure transducer senses the pressure drop. The nearly linear pressure-flow characteristic is limited to the laminar flow region, where the Reynolds numbers is less than 2000 (Fig 3–28). Although the pressure-flow relationship (resistance) appears linear in Figure 3–28, the conductance characteristic of the Fleisch flowmeter is nonlinear, as seen in Figure 3–29. Figure 3–30 shows that the addition of the screen upstream (Fig 3–27) causes important changes in the conductance characteristics of the Fleisch flowmeter.

As noted in Poiseuille's equation, the output of the Fleisch pneumotachograph is directly proportional to the viscosity of the flowing gas. These viscosity variations are gas composition and temperature dependent. For example, the resistance of the Fleisch can increase by 12.5% when the gas composition changes from room air (20.9% O_2) to 100% O_2. Figure 3–31 shows the conductance-flow characteristics for a Fleisch pneumotachometer with varying O_2–N_2 gas mixtures (from room air to 100% O_2). Figure 3–32 shows that the relative resistance with

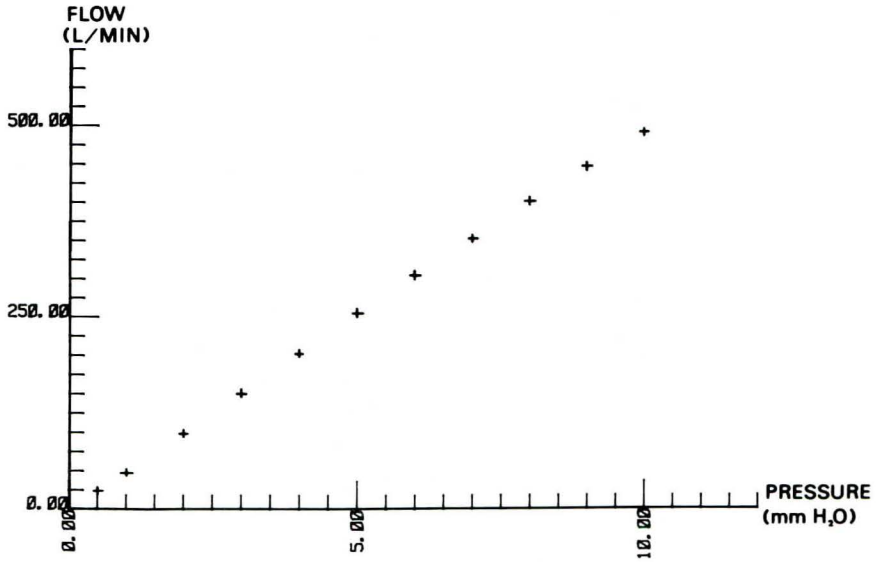

FIG 3–28.
Flow-pressure curve of a typical Fleisch No. 3 pneumotachometer provided by the manufacturer. (*From Yeh MP et al: Computerized determination of pneumotachometer characteristics using a calibrated syringe.* J Appl Physiol: Respir Envir Exercise Physiol *1982; 53:280–285. Used by permission.*)

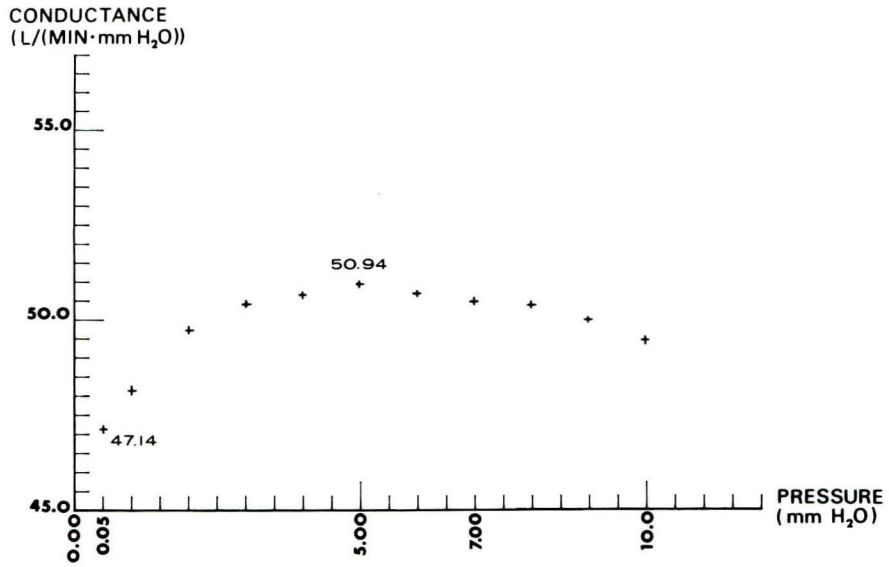

FIG 3–29.
Conductance-pressure curve for the same pneumotachometer as Figure 3–28. (*From Yeh MP et al: Computerized determination of pneumotachometer characteristics using a calibrated syringe.* J Appl Physiol: Respir Envir Exercise Physiol *1982; 53:280–285. Used by permission.*)

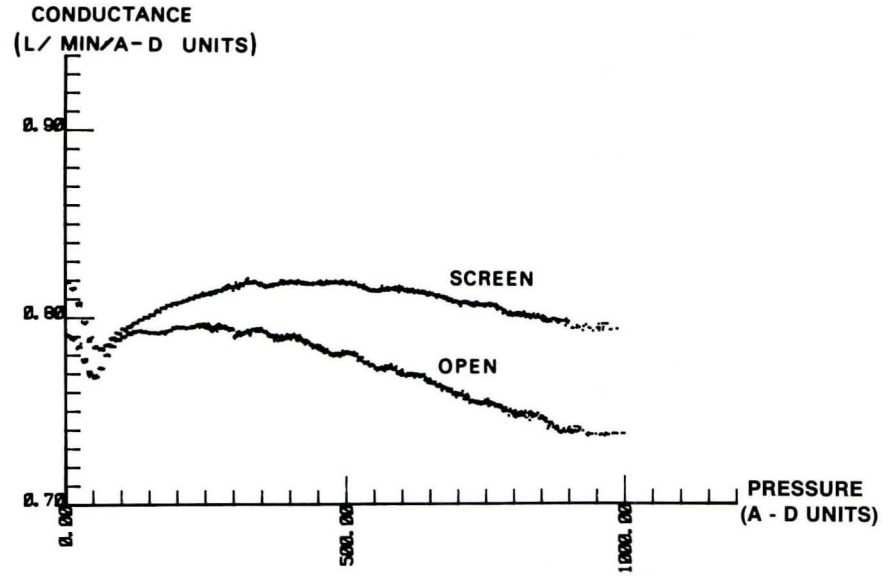

FIG 3–30.
Conductance characteristics of the same pneumotachometer with a wide-open upstream geometry (open curve) and with the manufacturer's upstream metal screen (screen curve). See Figure 3–27 for position of screen. (*From Yeh MP et al: Computerized determination of pneumotachometer characteristics using a calibrated syringe.* J Appl Physiol: Respir Envir Exercise Physiol *1982; 53:280–285. Used by permission.*)

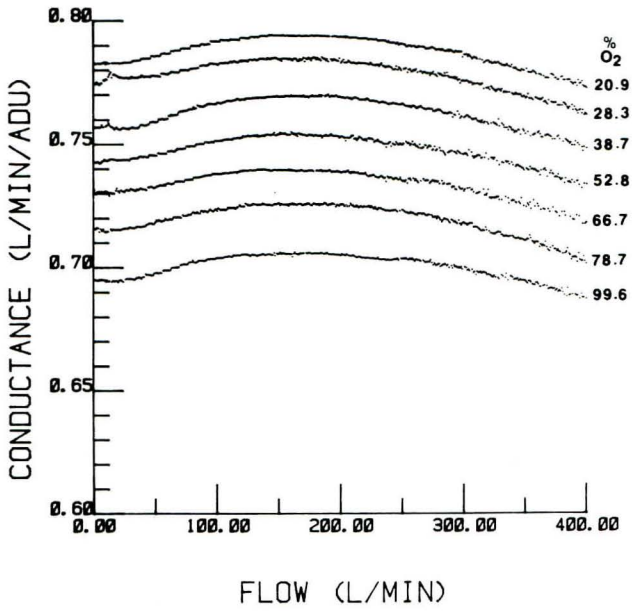

FIG 3–31.
Relative conductances with room air and O_2–N_2 gas mixtures. Note that the flow-conductance curves have the same shape. (*From Yeh MP et al: Effects of O_2, N_2 and CO_2 composition on nonlinearity of Fleisch pneumotachograph characteristics.* J Appl Physiol: Respir Envir Exercise Physiol *1984; 56:1423–1425. Used by permission.*)

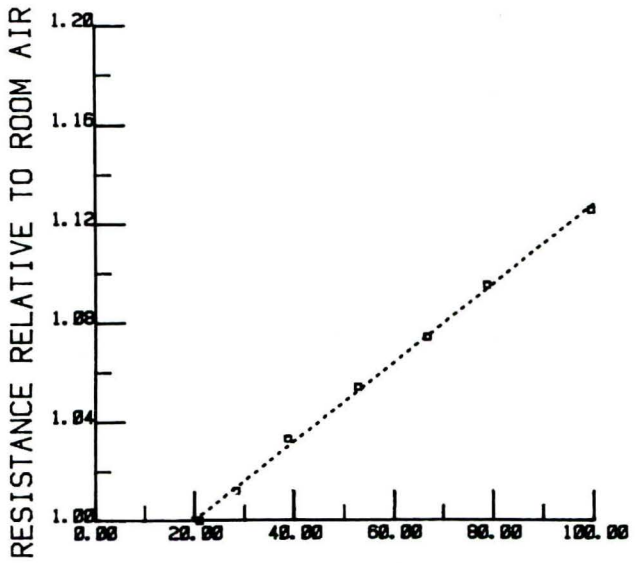

FIG 3–32.
Relative resistance of O2–N2 gas mixtures compare to room air. (*From Yeh MP et al: Effects of O_2, N_2 and CO_2 composition on nonlinearity of Fleisch pneumotachograph characteristics.* J Appl Physiol: Respir *Envir Exercise Physiol 1984; 56:1423–1425. Used by permission.)*

varying O_2 concentrations is linear. Yeh et al.[63, 64] have developed methods for accurately linearizing and calibrating Fleisch pneumotachographs to within about 1%.

Figure 3–33 shows a schematic diagram of a variable orifice flow-measuring device. The device has a nonlinear flow-pressure characteristic. However, the characteristic is highly reproducible. Because of its nonlinear flow-pressure characteristics, computerized "linearization" methods are used.[48]

The Wright respirometer is an example of a rotating-vane, or turbine, flow-measuring device. The flow of gas through the device turns the turbine at a rate dependent on the flow.[48] The movement of the turbine can be either mechanically or electronically coupled to a display device. If a mechanical mechanism is used, the turbine is linked to a mechanical display meter. If the rotations are sensed electronically, typically electronic pulses are generated by means of the turbine interrupting a beam of light. Each pulse is counted and is proportional to the flow. The inertia of the turbine makes it slow to respond and somewhat inaccurate for measuring breath-to-breath flow rates.

Thermal-element flowmeters use sensors such as thermistors and metal wires (hot-wire flowmeter) to sense flow.[48] The hot-wire anemometer uses a small heated element in the pathway of the gas flow. The current needed to maintain the element at constant temperature is measured and is proportional to the gas flow that cools the element. One limitation of most thermal-element flowmeters is that they are unidirectional. If measuring the direction of the flow is important, then directional valves and multiple flowmeters are used.

The two types of ultrasonic flowmeters are the transit-time and vortex-shedding types.[48] Figure 3–34 shows a diagram of the transit-time ultrasonic flowmeter. The time it takes the ultrasonic pulse to travel upstream and downstream changes, and it is based on gas flow, gas temperature, the speed of sound (c), the length of the pathway (L), and the angle of the sensors in relation to the stream as shown in these equations:

$$T_u = t_{upstream} = \frac{L}{c - v^*\cos(\beta)}$$

$$T_d = t_{downstream} = \frac{L}{c + v^*\cos(\beta)}$$

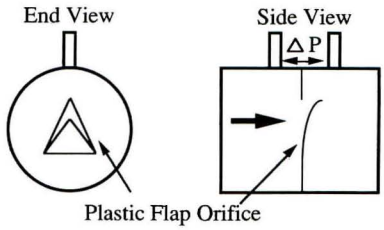

FIG 3–33.
Variable orifice flow transducer. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.)*

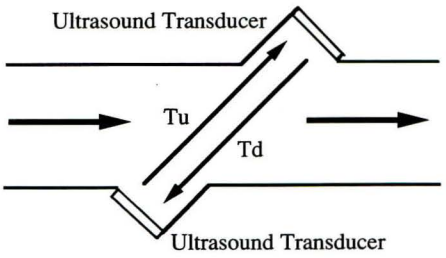

FIG 3–34.
Transit-time ultrasonic flowmeter. (*From East TD: What makes noninvasive monitoring tick? A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.)*

The vortex-shedding flowmeter (Fig 3–35) places obstructions in the airway so that whirlpools of gas (vortices) are produced.[48] The principle of vortex shedding has been known since 1878. The frequency at which vortices are shed is related to the flow of the gas past the obstruction. An ultrasonic transmitter is placed perpendicular to the flow. The ultrasonic signal is modulated by the frequency of the vortices, and flow is thus detected. The measurement is independent of gas density, viscosity, temperature, pressure, and conductivity. The design of the obstruction and the surrounding structures are primary factors that affect the calibration and linearity of this type of flowmeter. Vortex-shedding flowmeters are unidirectional.

Peak Flowmeters

Wright and McKerrow first reported the use of a simplified peak flowmeter in 1959—the Wright peak flowmeter.[65] Since that time this peak flowmeter has become the standard against which all other peak flowmeters have been tested and compared. In 1978 Wright described a simplified "mini-Wright" peak flowmeter.[66] This development and the competitive environment of modern medical instrumentation have lead to the proliferation of inexpensive peak

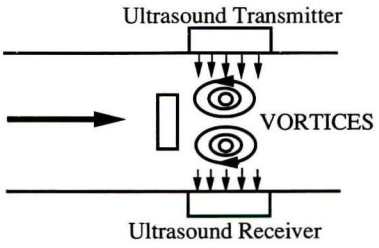

FIG 3–35.
Vortex-shedding ultrasonic flow transducer. (*From East TD: What makes noninvasive monitoring tick: A review of basic engineering principles.* Resp Care *1990; 35:500–519. Used by permission.)*

flowmeters. As a consequence, there have been a large number of reports in the literature comparing each new peak flowmeter with the original "Wright peak flowmeter," or espousing the advantage of the latest peak flowmeter.[67] There is some confusion about which is adequate or even best. As a result, the National Asthma Education Panel recently made recommendations on diagnosing asthma[68] and established technical standards and testing methods for peak flowmeters.[69]

Eight different models of peak flowmeters were recently tested using many different peak flow detecting technologies. The results of this testing showed—to the credit of the instrument manufacturers—that the quality, accuracy, and reproducibility of currently available peak flowmeters is surprisingly good.[67]

Gas Volume/Spirometers

Although the output of flowmeters can be integrated over time to obtain volume, several simpler and more accurate volume-measuring devices known as *spirometers* have been developed.[70–73] Basically, spirometers are calibrated containers that collect respiratory gases as they are exhaled. The simplest of these was used in the mid-1800s and was simply an inverted cylindrical "bell" that used a water bath as a seal. The vertical displacement of the bell and its known cross-sectional area were used to measure volume. More recently, recorders were added to spirometers with a time base so that the more dynamic characteristics of volume and flow could be easily determined. The American Thoracic Society (ATS) has taken the leadership to standardize spirometers used in pulmonary function laboratories.[72]

Spirometers are devised using a variety of technologies; a water seal, as described above; a rolling seal, in which a cylinder moves horizontally with a very thin (usually plastic) seal; a wedge bellows; and a variety of flow sensors. Recently, the characteristics of a wide variety of spirometers have been measured.[73] Most, but not all, commercially available spirometers meet the ATS requirements.

Plethysmography

Plethysmographs measure volume. When it is impossible or inconvenient to measure the volume of body parts such as the gas in the lungs or blood in the lower leg, plethysmographs are used. For measurement of thoracic gas volume, a body plethysmograph is frequently used.[70, 71] For a volume of gas at constant temperature, the product of pressure (P) and volume (V) is a constant (Boyle's law). The patient is seated in a body plethysmograph, with the nose occluded, breathing quietly, through a tube to the outside. When the subject is breathing at functional residual capacity (FRC) the airway is closed, trapping the gas in the lung at near FRC. The subject then gently pants against a closed shutter at about two times per second. The pressure at the mouth is measured as is the volume change in the body box, which is measured via either a plethysmograph or a spirometer. The pressure and volume signals measured with the appropriate transducers are then applied to allow measurement of lung volume. Plethysmography for the legs and other body parts applies the same transducers and principles.

Biosensors and Electrochemical Sensors

Biosensors are sophisticated descendants of the canary in the coal mine.[74, 75] Modern biosensors have evolved as a result of two disparate disciplines: information technology with its microcomputers and optical fibers, and molecular biology. With such sensors, faster measurements can now be made at the bedside at potentially less expense than with current clinical laboratory measurements. Leland C. Clark, Jr., inventor of the oxygen electrode, is credited with developing the first biosensor.[74] Clark extended his "oxygen electrode" to measure blood glucose by coating his electrode with a gel containing a biocatalyst. Currently, roughly 100 different enzymes are used in biosensors, and the future points to an ever-increasing number. In the future, miniature disposable implantable biosensors, and eventually noninvasive biosensors, may become available to detect many of the body's chemical and gas measurements.

VENTILATORS: AN INTEGRATED INSTRUMENTATION COMPLEX

The modern ventilator is a place where many of the transducers discussed in this chapter come together. Today's ventilator is a complex unit of sensors, electronics, microcomputers, and software engineering.[76] These ventilators can be modified easily by a simple software change to provide almost any new mode of ventilation and to collect information from a variety of sensors. This information can be displayed, stored, and manipulated in a variety of ways—limited only by the imagination of software engineers. Ventilators of the future will likely have plug-in modules to perform pulse oximetry, breath-by-breath infrared CO_2 analysis, mixed venous oximetry, and perhaps even noninvasive blood pressure monitoring. Communications capabilities to provide convenient and practical electronic interfaces with other monitoring devices will become common.

Computer-aided protocols may eventually find their way into ventilators.[76] The need for better and more consistent respiratory monitoring has been pointed out by many investigators.[76–81] Before these computer protocols can be applied, however, dramatic improvements must occur in the automatic acquisition processing and recording of data.[76] Dealing with only the "raw" signals coming from the transducers, with the "artifacts/noise" and the complexity of signals, is likely to stymie progress with standardized computerized protocols. The transducers we currently have available to us in respiratory monitoring are sufficiently robust, accurate, and stable at present to allow computer-driven protocols to be developed. However, understanding the complexity of signal versus noise will be a difficult task—one that will require the efforts of the larger pulmonary community to solve. Such an activity will require careful review of the problems and the development of standard methodologies so that a medical consensus can be achieved. The future for respiratory care monitoring is bright and the challenges achievable. We must now develop a strategy and "team" to improve the quality of monitoring, not just the quantity of signals we measure.

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